DEVELOPMENT AND INVESTIGATION OF SINGLE-SCAN TV RADIOGRAPHY FOR THE ACQUISITION OF DYNAMIC PHYSIOLOGIC DATA

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RESEARCH SUMMARY

1. Localization and Tracking of Implanted Screws.

In order to facilitate the processing of either ciné roentgenograms or on-line processing of video disc recorded images we have devised a simple algorithm for localization of implanted screws or other markers attached to moving segments of organ anatomies. This program retains all the features discussed in our last semi-annual report but substitutes the placing of an electronic window over the area for the computer operation. This window of appropriate size can be loosely positioned and therefore relieves the tedious and time consuming task of placing a cross-hair in a precise position.

The program is capable of locating a small area having varying degrees of contrast with respect to its surroundings. The electronic window simply acts as an instruction for the computer, telling it to confine its search to the area enclosed by the window. By having this step of operator intervention we eliminate the major obstacle to automatic localization by the computer. The first step in the procedure is the digitization of the area enclosed by the window. In the examples to be shown we have used an area of 25 x 25 picture elements. Each element represents in a normal TV image 1/240 vertically and 1/320 horizontally of the entire image. In a high resolution system this can be expanded to 1/1000 of the entire image size. The digitization if accomplished using the method reported by Baily, and Crepeau,[1] into 32 gray levels. In the case of screws placed on the inner wall of the heart, the dimensions of the screws would be of the order of 3 or 4 picture elements.

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1. Capabilities of a Single Scan TV-Radiographic System for Digital Data Acquisition. N.A. Baily and R.L. Crepeau, Invest. Radiol. 6, 273-279 (1971).

The area surrounding the object to be localized does not have to be of uniform brightness and will of course contain noise components in the form of quantum mottle and electronic noise generated in the recording, playback, and digitization stages. In addition non-uniformities due to the x-ray tube and image amplifier characteristics are usually present.

The two step algorithm which was devised for this localization procedure is insensitive to either the presence of these perturbations or their characteristics. The two steps are: (1) Preprocessing, and (2) Thresholding.

A. Preprocessing.

Since the preprocessing is a completely computerized operation and depends upon an examination of each point in the array with respect to its surroundings, if the objects to be localized were too close to the border of the electronic window it would be missed. In the present program this distance must be greater than 3 picture elements or approximately 10 percent of the window size from the edge of the window. This should not pose any problems for the operator.

Each point, M_{ij}, is then considered to be the center of a 7 x 7 array and also the center of a smaller array of picture elements having dimensions of 3 x 3 elements. The average gray scale value is then obtained for the central 9 elements, $A(9)_{ij}$, and also for the remaining 40 picture elements in the 7 x 7 array, $A(40)_{ij}$. The normalized contrast between $A(9)_{ij}$ and $A(40)_{ij}$ is then calculated and assigned to each M_{ij}

$$C_{ij} = \frac{100}{3} \frac{[A(9)_{ij} - A(40)_{ij}]}{A(9)_{ij}}$$

for situations where the average gray level of $A(9)_{ij}$ is numerically greater than that of $A(40)_{ij}$.

In the case of a screw presenting as a bright spot in the image the value of C_{ij} will be large for all M_{ij} residing within the image of the screw. If M_{ij} represents a picture element that does not represent either a portion or the entire image of a screw, C_{ij} will be small or possibly even negative.

The original array consisting of the values M_{ij} is now replaced by a new array, C_{ij}, where C'_{ij} is;

$$C'_{ij} = \begin{cases} C_{ij}, & \text{when } C_{ij} \gtrsim 0 \\ 0, & \text{when } C_{ij} < 0 \end{cases}$$

As mentioned above, this procedure eliminates all points in a 3 element band on the outside of the original array leaving C_{ij} as a 19 x 19 array, or for any square array, M_{ij}, of dimension D, the C_{ij} array will have dimensions equal to D-6.

B. Thresholding.

This final step simply consists of a determination of the center of mass of the values of the C'_{ij} matrix which remain after values below a selected value have been eliminated from the matrix. Data acquired for determining a method to select this value will be presented in a later section.

The choice of array sizes is also variable. The values chosen are influenced by the size of the screws (3-4 picture elements) in relation to the size of the left ventricle as imaged on a typical ciné radiographic unit. This choice provides discrimination against larger bright objects since such images will have portions both in the 3 x 3 and in the surrounding matrix thereby making the computed contrast, C_{ij}, much lower than if the image to be located was substantially within the 3 x 3 array.

The flow charts for the two steps in the computer program are shown in Figs. 1 and 2.

C. Testing.

Noise incorporated in the image, particularly if it appears as spikes superimposed on the video signal, can generate errors in the computation of the position of the object to be localized. It has been shown,[2] that at least a large

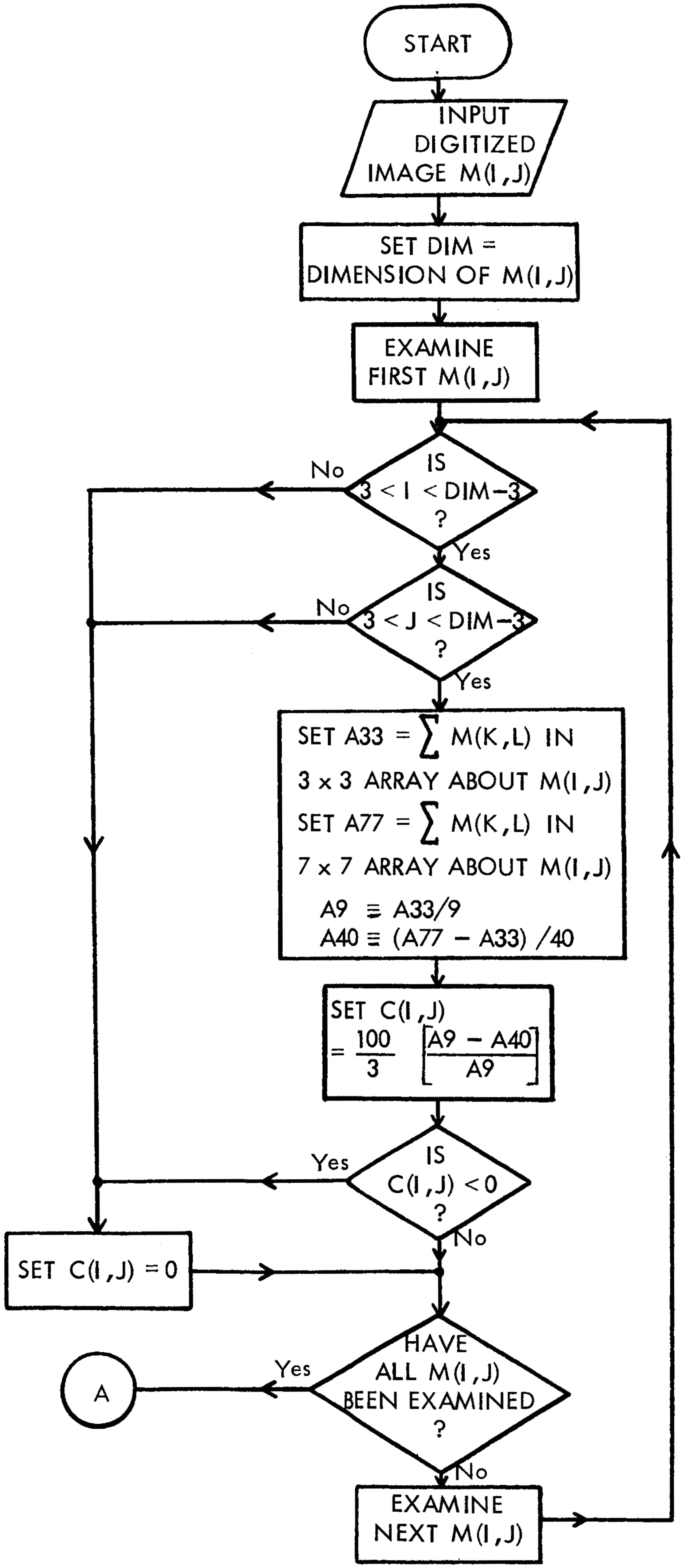
START
INPUT DIGITIZED IMAGE M(I,J)
SET DIM = DIMENSION OF M(I,J)
EXAMINE FIRST M(I,J)
IS 3 < I < DIM-3 ?
No
Yes
IS 3 < J < DIM-3 ?
No
Yes
SET A33 = ∑ M(K,L) IN 3 x 3 ARRAY ABOUT M(I,J)
SET A77 = ∑ M(K,L) IN 7 x 7 ARRAY ABOUT M(I,J)
A9 ≡ A33/9
A40 ≡ (A77 − A33) /40
SET C(I,J) = 100/3 [(A9 − A40)/A9]
IS C(I,J) < 0 ?
Yes
No
SET C(I,J) = 0
HAVE ALL M(I,J) BEEN EXAMINED ?
Yes
No
A
EXAMINE NEXT M(I,J)

FIGURE 1

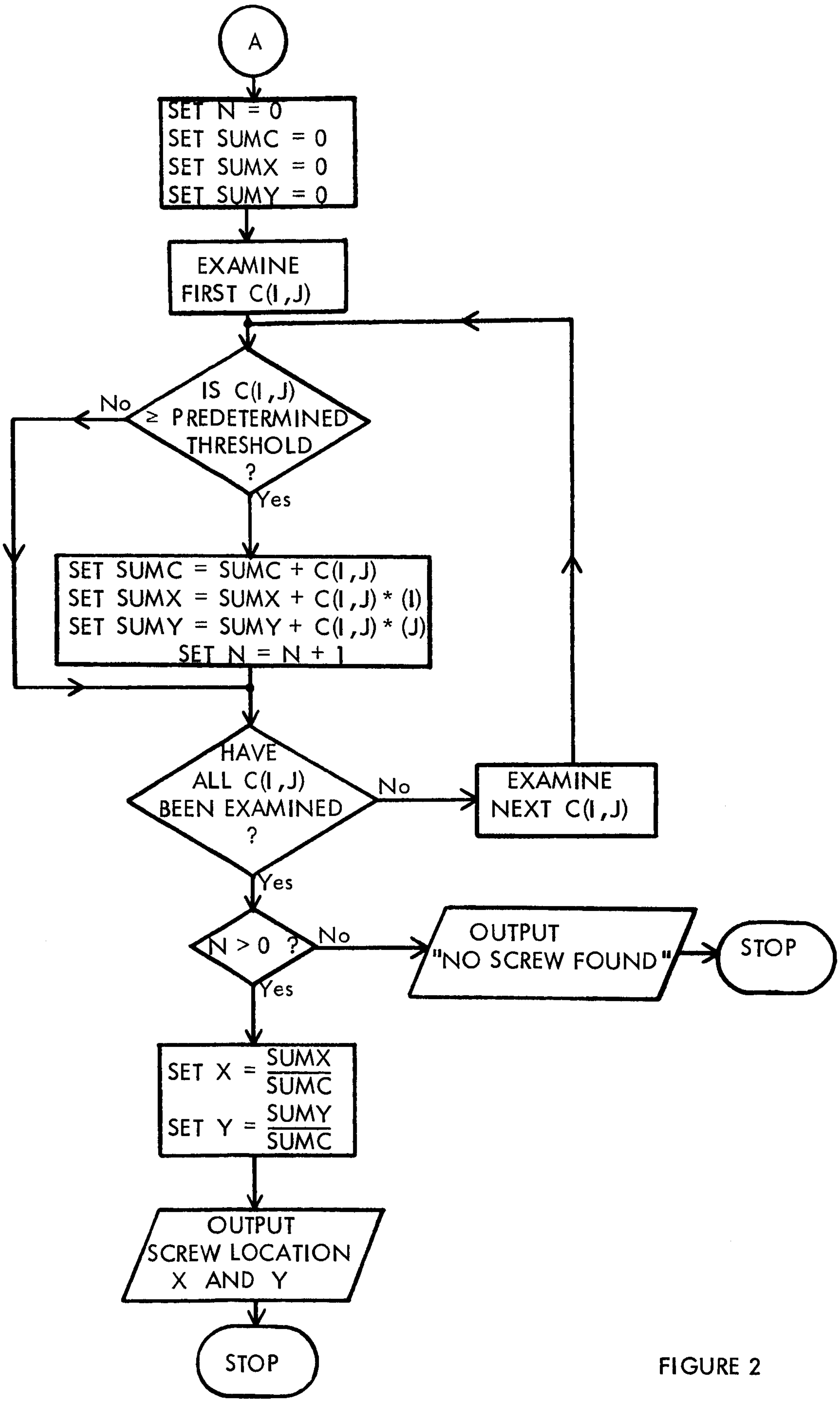
A
SET N = 0
SET SUMC = 0
SET SUMX = 0
SET SUMY = 0
EXAMINE FIRST C(I,J)
IS C(I,J) ≥ PREDETERMINED THRESHOLD ?
No
Yes
SET SUMC = SUMC + C(I,J)
SET SUMX = SUMX + C(I,J) * (I)
SET SUMY = SUMY + C(I,J) * (J)
SET N = N + 1
HAVE ALL C(I,J) BEEN EXAMINED ?
No
EXAMINE NEXT C(I,J)
Yes
N > 0 ?
No
OUTPUT "NO SCREW FOUND"
STOP
Yes
SET X = SUMX / SUMC
SET Y = SUMY / SUMC
OUTPUT SCREW LOCATION X AND Y
STOP

FIGURE 2

portion of the noise is statistically independent. In such cases spatial averaging of the data, combined with temporal averaging will diminish its effect. This was accomplished in two ways. First, multiple digitizations were averaged to provide the initial array. Second, averaging (spatial) is involved in arriving at values for both $A(9)_{ij}$ and $A(40)_{ij}$.

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2. Detection capability of differential radiographic absorption through the use of digitalized fluoroscopic images. R.L. Crepeau and N.A. Baily, Proc. of the San Diego Biomed. Symp. 1973. 12, 277-283 (1973).

In order to test the program and establish a positive method for choosing the proper parameters, four 25 x 25 digitized test patterns, representing typical cardiac situations, were punched on paper tape. Each of these was characterized by a screw set off-center about half way along a diagonal of the square, having dimensions of 3 x 3 picture elements, and differing in brightness from a uniform background of 1, 2, 4, and 6 gray levels (out of 32). These tapes were then processed and screw locations determined for all possible threshold values. Integer values of 1, 2, . . . $C'_{ij}\big|_{Max}$ were used successivley.

These patterns were then modified by the addition of random noise patterns having a large range of σ's (standard deviation). In a random noise pattern the σ is equivalent to the RMS value of the noise. The video signal-to-noise level is usually taken to be 6 x RMS.

In the test program each digitized picture element having an integer value between 0 - 31 in the noise-free pattern, M_{ij}, was replaced by a pseudo continuous value placed at the center of the level. That is a value at point $M_{ij} = 8$ would now have the value 8.50000 . . . . A value of noise N_{ij} arrived at from a sampling of a random noise generator was then added to each of the new M_{ij} values. The random noise generator produced numbers lying in the range of continuous gray levels and having a mean value of zero and a standard deviation σ in gray scale units. The sum of each $M_{ij} + N_{ij}$ was then rounded off to the nearest integer, $\overline{M}_{ij}$, and formed into a new array closely representing the type of arrays arrived at by digitization of TV roentgenographic or fluoroscopic images. The signal-to-noise level is specified by the standard deviation of the added noise and the difference in gray level of the object and background. These were then subjected to the algorithm described above. The displacement of the coordinates from the noise-free coordinates found for the screw position in each case was taken as a measure of the error introduced by various degrees of noise. The results are shown in Tables 1 through 4. This data has not been completely analyzed but it is expected that this data will furnish a family of curves which will allow quick selection of the proper parameters for the processing of any particular set of images.

2. Contour Identification.

In our last report we described a method using an operator positioned set of cross-hairs (on a TV monitor) for deriving contours of either the inner or outer walls of the cardiac chambers. We have now developed a computer algorithm which is capable of developing such contours automatically. It is therefore possible using this new technique to generate such contours on-line from successive video disc recordings.

Both in single radiological images and in time sequences of such images a great deal of quantitative data is incorporated as changes of density or intensity. These changes are indicative of physiological change and/or functions derivable from size and/or shape of these structures. Dynamic information is obtainable from time sequences of such images. A first step in extracting such information in most cases is to provide the computer with information concerning the boundaries. In the case of single images, both areas and volumes can be obtained. In the case of time sequences of such images, changes in these quantities as a function of time can be computed and plotted automatically. Velocity and acceleration of the boundaries are readily obtained from such sequences, and analysis of the frequency components of such motions can also be readily computed. In all cases, the accuracy, and indeed the availability of such data depends on the ability to

TABLE I

SCREW DISPLACEMENT (PICTURE ELEMENTS)

Gray Level Difference = 6

THRESHOLD =	1		2		3		4		5		6		7		8		9		10		11		12		13	
σ	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY
0.2	0.04	0.06	0.04	0.06	0.04	0.01	0.04	0.09	0.06	0.11	0.18	0.01	0.08	0.03	0	0.11	0.2	0.03	0.02	0.15	0.26	0.49				
0.3	0.21	0.83	0.06	0.17	0.02	0.01	0.08	0.05	0.04	0.09	0.14	0.05	0.14	0.07	0	0.15	0.16	0.09	0.02	0.13	0.26	0.18				
0.4	2.2	2.07	0.68	0.95	0.02	0.05	0.04	0.01	0.06	0.09	0.18	0.07	0.14	0.03	0.04	0.09	0.1	0.03	0	0.01	0.26	0.01				
0.5	3.22	3.25	1.88	1.91	0.46	1.29	0	0.25	0.1	0.03	0.16	0.09	0.12	0.07	0.02	0.05	0.04	0.05	0.26	0.01	0.26	0.01				
0.6	3.82	3.71	2.84	2.57	1.46	1.39	0.3	0.39	0.1	0.03	0.16	0.05	0.16	0.07	0	0.09	0.04	0.13	0.085	0.065	0.26	0.26	0.26	0.49		
0.7	4.34	4.13	3.56	3.25	2.46	2.29	0.84	1.05	0.08	0.51	0.12	0.25	0.1	0.03	0.02	0.09	0.06	0.03	0.235	0.06	0.26	0.01	0.26	0.01		
0.8	4.8	4.63	4.18	1.97	3.22	3.13	1.78	1.83	0.58	1.17	0.06	0.49	0.06	0.21	0.06	0.11	0.1	0.19	0.26	0.015	0.26	0.24				
0.9	5.04	4.93	4.7	4.47	3.88	3.79	2.48	2.71	1.08	1.57	0.04	0.75	0.04	0.49	0.1	0.03	0.04	0.15	0.235	0.06	0.26	0.24	0.26	0.51		
1.0	5.18	5.07	4.86	4.87	4.26	4.27	3.12	3.33	1.44	1.79	0.46	0.87	0.02	0.59	0.04	0.03	0.02	0.09	0.185	0.11	0.41	0.14				
1.1	5.5	5.29	5.34	5.13	5.02	4.83	4.26	3.95	2.76	2.95	1.98	2.19	0.44	1.03	0	1.15	0.16	0.05	0.31	0.01	0.41	0.11	0.26	0.51		
1.2	5.56	5.45	5.44	5.33	5.24	5.05	4.56	4.77	3.46	3.43	2.36	2.95	1.18	1.79	0.08	1.25	0.14	1.05	0.135	0.065	0.41	0.39	0.26	0.01		
1.3	5.6	5.5	5.6	5.3	5.4	5.1	4.9	4.9	4.3	4.2	2.7	3.1	1.9	2.4	0.2	1.4	0.1	1.7	0.1	0.1	0.3	0.1	0.5	0.0		
1.4	5.8	5.6	5.7	5.5	5.6	5.4	5.3	5.4	4.8	4.8	3.9	3.7	2.6	3.0	1.9	2.4	0.0	1.5	0.2	3.4	0.3	0.0	0.5	0.0		
1.5	5.9	5.8	5.8	5.7	5.7	5.6	5.5	5.5	4.9	5.2	4.2	4.6	3.0	4.0	2.4	3.6	1.9	2.6	0.4	3.1	0.1	6.1	0.5	0.0	1.3	0.5
1.6	6.0	5.8	6.0	5.8	5.8	5.7	5.7	5.7	5.3	5.4	5.0	5.0	3.7	4.2	2.6	3.7	2.1	3.4	0.2	2.8	0.3	6.0	0.1	6.3	1.3	0.5

TABLE II

SCREW DISPLACEMENT (PICTURE ELEMENTS)

Gray Level Difference = 4

THRESHOLD =	1		2		3		4		5		6		7		8		9		10	
σ	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY
0.2	0.1	0.0	0.0	0.0	0.0	0.0	0.2	0.0	0.1	0.1	0.2	0.1	0.1	0.2						
0.3	0.5	0.8	0.0	0.0	0.0	0.0	0.2	0.0	0.1	0.0	0.1	0.1	0.0	0.2						
0.4	2.6	2.3	0.5	0.8	0.1	0.1	0.2	0.1	0.1	0.0	0.0	0.1	0.2	0.0	0.3	0.0				
0.5	3.6	3.7	1.7	2.1	0.2	0.7	0.2	0.0	0.1	0.0	0.0	0.0	0.4	0.1	0.3	0.0				
0.6	4.2	4.1	2.8	2.8	0.8	1.1	0.1	0.4	0.2	0.1	0.0	0.2	0.3	0.1	0.3	0.0				
0.7	4.7	4.5	3.7	3.5	2.2	2.3	0.3	0.7	0.2	0.3	0.2	0.1	0.3	0.1	0.2	0.1	0.3	0.0		
0.8	5.2	5.0	4.4	4.3	3.2	3.3	1.4	2.0	0.1	0.9	0.1	1.0	0.3	0.2	0.3	0.1				
0.9	5.2	5.3	4.9	4.9	3.8	4.1	2.1	3.1	0.6	1.6	0.1	1.3	0.3	0.1	0.4	0.2	0.3	0.0		
1.0	5.5	5.5	5.1	5.2	4.4	4.8	3.1	3.4	1.1	1.8	0.3	1.5	0.1	0.4	0.5	0.1	1.3	0.5		
1.1	5.8	5.5	5.7	5.4	5.3	5.0	4.2	4.1	2.8	3.2	0.9	2.0	0.3	1.9	0.3	0.1	0.8	0.5		
1.2	5.8	5.8	5.7	5.6	5.5	5.5	4.9	5.1	3.2	4.0	2.4	3.5	0.8	2.3	0.3	5.8	0.5	0.2		
1.3	6.0	5.8	6.0	5.6	5.7	5.5	5.3	5.4	4.4	4.5	3.1	3.8	1.5	2.4	0.3	6.0	0.5	0.2	1.3	0.5

TABLE III

SCREW DISPLACEMENT (PICTURE ELEMENTS)

Gray Level Difference = 3

THRESHOLD =	1		2		3		4		5	
σ	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY
0.2	0.0	0.0	0.1	0.0	0.2	0.1	0.2	0.1		
0.3	0.5	1.1	0.1	0.1	0.1	0.1	0.2	0.1		
0.4	3.4	3.1	0.1	0.6	0.1	0.3	0.1	0.3		
0.5	4.8	4.5	2.5	2.7	1.1	1.1	0.4	0.1		
0.6	5.5	4.8	4.3	3.7	1.2	1.8	0.5	0.2	0.7	0.2
0.7	5.8	5.4	5.2	4.6	3.1	3.3	0.5	6.1	0.0	4.9

TABLE IV

SCREW DISPLACEMENT (PICTURE ELEMENTS)

Gray Level Difference = 1

THRESHOLD =	1		2		3	
σ	ΔX	ΔY	ΔX	ΔY	ΔX	ΔY
0.1	0.0	0.0	0.0	0.0		
0.2	0.0	0.2	0.1	0.4		
0.3	0.6	1.3	0.2	0.3		
0.4	4.2	3.8	0.2	0.6	0.8	1.3
0.5	5.5	5.1	4.0	4.2	1.4	3.3

extract contours with good accuracy. Trenholm, et al,[3] have shown that when a contour has been defined and the gray scale level of each point within the contour is known, accurate values for volumes can also be easily obtained. Using digitized values and some combination of both time and spatial averaging to increase the signal-to-noise ratio, even higher accuracy should result since small gray scale differences not detectable by analog techniques will now be meaningful.[2] This ability combined with an accurate contour defining the outline in a single plane should produce highly accurate volumetric data.

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3. B.G. Trenholm, D.A. Winter, G.D. Reimer, D. Mymin, E.L. Lansdown, and G.P. Sharma, Radiology 112, 299 (1974).

In general, the method to be described works best where high contrasts exist. However, since even a minimal amount of statistical averaging provides a value having a standard deviation in the determination of a mean gray level of $\pm$ 1 percent or approximately $\pm$ 0.3 gray levels, we feel that the method should certainly be useful for structures outlined by changes of gray level corresponding to about twice this value.[2]

METHOD

The basic method used in the work to be presented is based on two techniques presented in the book by Rosenfeld.[4] These are a calculation of the gradient of each point in our digitized image and a thresholding technique designed to eliminate all values of the gradient not having a sufficient differential value with respect to its nearest neighbors.

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4. Picture Processing by Computer, A. Rosenfeld, Academic Press, N.Y. (1969).

The first step in the process is the digitization of the image or section of image to be contoured. In other words we produce a picture function $f(x, y)$ having a range of values from 0 - 31. These 32 gray levels correspond to the range of fluoroscopic brightness or roentgenographic densities found in the image to be processed. Various techniques for eliminating statistically independent electronic noise such as spatial averaging,[2] can be employed to yield values for $f(x, y)$ having a high degree of reliability especially with respect to their relative values. The second step involves a numerical calculation to obtain the gradient of $f(x, y)$ for all values of x and y. The directional derivative of each point was calculated using the eight closest neighbors to the point of calculation. For example, given a series of nine picture elements arranged in the following manner:

$$
\begin{matrix}
A_{x-1,\, y+1} & A_{x,\, y+1} & A_{x+1,\, y+1} \\
A_{x-1,\, y} & A_{x,\, y} & A_{x+1,\, y} \\
A_{x-1,\, y-1} & A_{x,\, y-1} & A_{x+1,\, y-1}
\end{matrix}
$$

the gradient of $A_{x,y}$ was computed in the following manner,

$$\left|\overrightarrow{\text{GRAD}}\,(A_{x,y})\right| = 1/3\ \Big[\,\big|(A_{x-1,y+1} + A_{x,y+1} + A_{x+1,y+1}) - (A_{x-1,y-1} + A_{x,y-1} + A_{x+1,y-1})\big| + \big|(A_{x-1,y+1} + A_{x-1,y} + A_{x-1,y-1}) - (A_{x+1,y+1} + A_{x+1,y} + A_{x+1,y-1})\big|\,\Big]$$

To obtain a contour, this matrix of values of the gradient of the original picture function is examined for a selected minimum value. This operation is similar to that usually called thresholding which is employed to make features of greater brightness stand out or even for outlining areas. In our case the threshold is applied to the values of the gradient rather than the values of $f(x, y)$. In the examples we will show, the threshold value for the gradient which we applied was equivalent to three gray levels. That is, if the gradient of a picture element with respect to its nearest neighbors did not exceed 10 percent of the gray scale range its value was set equal to zero. This procedure produces contours which in general are thicker than one would desire. To improve the resolution one must narrow down the number of picture elements still retained in the gradient matrix.

In order to eliminate noise spikes, artifacts, etc. from the gradient matrix of the picture elements the following thinning algorithm was used. Each value of the gradient function $g(x, y)$ is treated as the center point of a 3×3 array. Each array is examined and each value of $g(x, y)$ that has less than two non-zero values in the 3×3 array is set equal to zero. This is then followed by zeroing all points which have four or more nearest neighbors who themselves have more non-zero neighbors than the point being examined. This algorithm is applied twice and then followed by a gradient maximum following algorithm designed to eliminate all but the greatest values of the gradients remaining in the matrix.

The gradient maximum following algorithm operates in the following way. Each remaining point in the $g(x, y)$ matrix of the gradient of the picture elements is considered as the center of a 3×3 array of its nearest neighbors as done earlier. The average value of the nine gradients is compared to the value of the gradient under consideration and if the average value exceeds the gradient then the point is zeroed. The derived contour then is just the points remaining with non-zero values in the matrix representing $g(x, y)$.

RESULTS

The contouring method described above has been tested on a number of different anatomical structures such as the heart, kidney, brain, and gallbladder. Two of these will be illustrated below.

Figure 3 is an example of a cardiovascular application. Since no appropriate roentgenogram was available for our use, recourse was made to the published literature.[5]

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5. M.G. Baron, Radiol. Clinics of No. America <u>6</u>, 353 (1968).

Figure 3A is a selective left ventricular angiogram, lateral projection, and shows a ventricular septal defect. To obtain this picture a negative of the illustration given in reference 5 was used. B, is the computer-generated contour. It is important to note that the program has even delineated contours of relatively low contrast as well as those of relatively high contrast.

Figure 4 is an example of the method's use in a genito-urinary application. A, is a print made from the original roentgenogram obtained during selective arteriography of the left kidney, and B, is the resultant computer-generated contour.

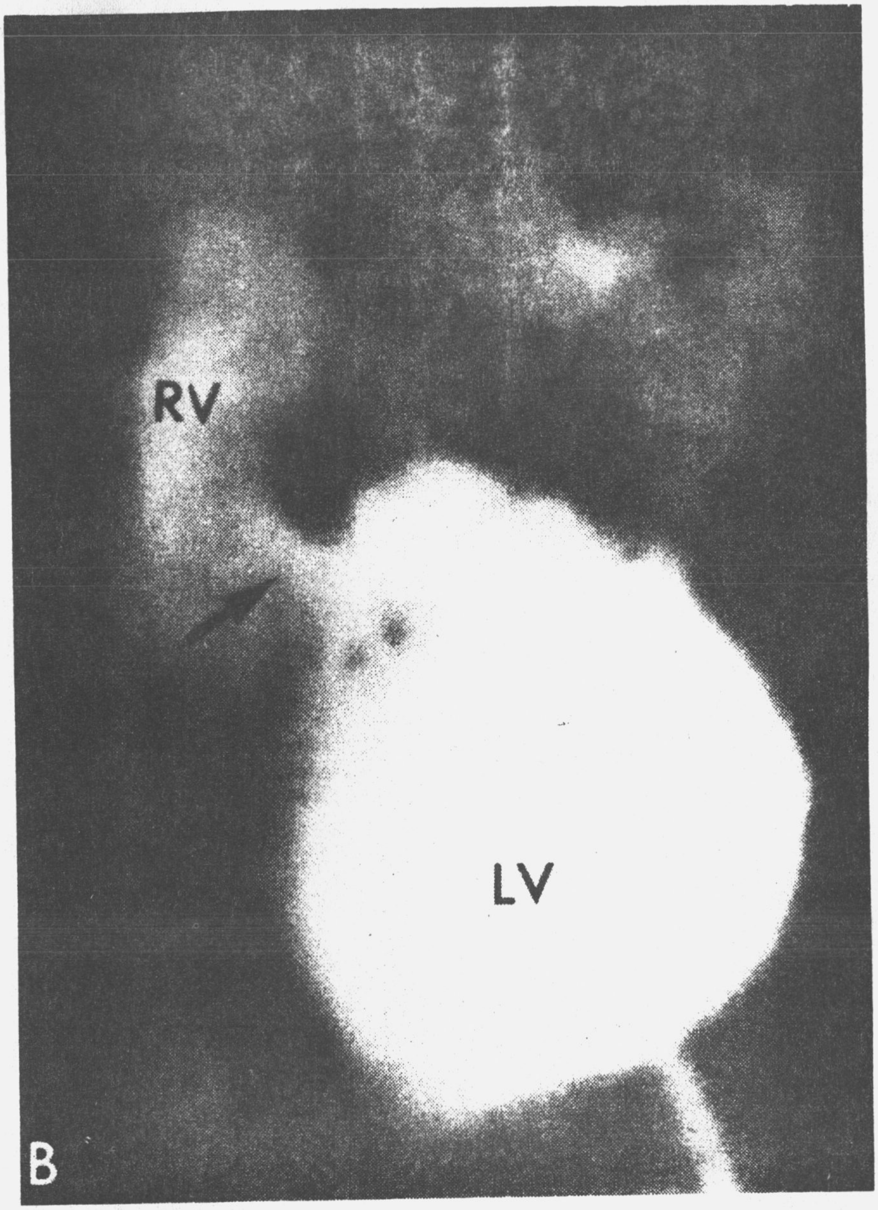

FIGURE 3A

FIGURE 3B

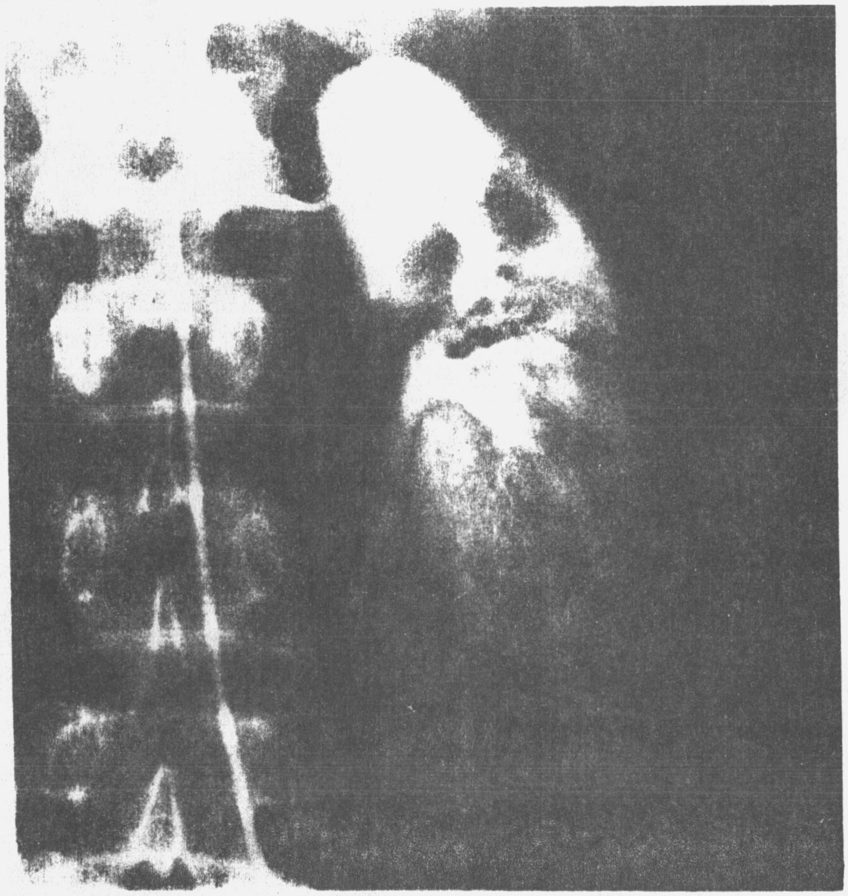

FIGURE 4A

FIGURE 4B

DISCUSSION

The simple method presented for extracting contours representing density changes of roentgenograms or fluoroscopic brightness changes should be useful for extracting physiological data in the form of organ areas and volumes. The digital data residing in the computer is easily utilized for computations of enclosed areas or the calculation of dynamic changes. Extraneous areas such as those produced by catheters, bony structures, etc. can be eliminated by operator intervention before beginning the computation of such areas. Since gray scale information on each of the picture elements enclosed is also stored in the computer, volumes of such structures can be calculated using the method of Trenholm, et al.[3]

The method is superior to simple thresholding of gray levels as described by Robb[6] in that the use of a gradient function allows one to operate with lower contrast levels. It is also faster and less tedious than the method used by Hall, et al,[7] which again basically uses simple thresholding techniques.

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6. R.A. Robb, Proc. of the San Diego Biomed. Symp. 10, 235 (1971).
7. D.L. Hall, G.S. Tadwick, R.P. Kruger, S.J. Dwyer, and J.R. Townes, Radiology, 101, 491 (1971).

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3. Metrology.

The experiments designed to measure the precision and accuracy of the linear dimensions of structures incorporated within an image showed an absolute standard deviation of ± 10%. Since the precision of such measurements was of the order of 0.5%, the absolute error can be attributed to circuit deficiencies. This was found to be the case in that the DEC-ADC's have an uncertainty in reproduction of the last bit. To overcome this we have developed a new electronic system based on a crystal oscillator with a precision of ± 0.1%. The test experiments will now be repeated to determine the system's capability to make accurate absolute measurements. We have also designed this new unit to remove the small amount of non-linearity found when small objects were being measured.

4. Cooperative Project for Use of a Digitized Fluoroscopic X-Ray System On The NASA Ames Centrifuge.

We have furnished schematics, systems diagrams, and consulted with Ames personnel with the aim of having a completely operating system sometime during the month of November in preparation for an animal run scheduled for shortly after the first of the year. We expect to provide aid in assembly, testing, and operation of this new x-ray system. This system will incorporate the large screen fluoroscopic unit described in the last report.

5. Processing of Ames Laboratory Data.

The equipment required for processing ciné angiographic studies or fluoroscopically recorded images requires essentially the same equipment as that which will be used on the centrifuge. We will assist the Ames personnel in transferring the computer programs and in assembling this equipment into an operating system. This system will make use of only standard commercial components.

6. Computerized Reconstruction of Images From Fluoroscopic Inputs.

In a basic and elegant application of the digital computer to the field of radiology, Hounsfield of E.M.I. Research Laboratories in the U.K. has used the computer to unravel the three-dimensional information previously superimposed in two dimensions on a normal radiograph. The digitized information is obtained by rapid transverse axial scanning of a collimated x-ray beam and Na I detector combination. The result is a transverse axial section where areas having contrast differences of about three percent and dimensions of the order of a centimeter can be resolved from their surroundings. The method is presented in detail by Cho, et al.[8] Some clinical investigations have also been presented recently by Ledley, et al.[9] A simpler, faster, cheaper, more available, and higher resolution system would result if similar results could be achieved using images recorded by an ordinary clinical type fluoroscopic unit. In conjunction with Dr. A. Kak of Purdue University we have succeeded in producing such sections from fluoroscopically recorded images.

8. Z.H. Cho, I.Ahn, C. Bohm, and G. Huth, Phys. Med. Biol. 19, 511 (1974).

9. R.S. Ledley, G. DiChiro, A.J. Luessenkop, and H.L. Twigg, Science 186, 207 (1974).

Using a phantom consisting of a luctie block mounted on a 1 mm sheet of aluminum, 60 images recorded 3° apart were made. The lucite block had dimensions of 2 cm x 3 cm in cross-section. Two holes, each 1 cm in diameter, were drilled and these filled with a 2.5% solution of Hypaque. The results are shown in Figs. 5 and 6. Both were made from the same reconstruction with the display set at different levels. As can be easily seen, even the 1 mm thick Al sheet is well resolved. The various exposures shown in each figure represent different photographic exposures.

PAPERS PRESENTED AT MEETINGS

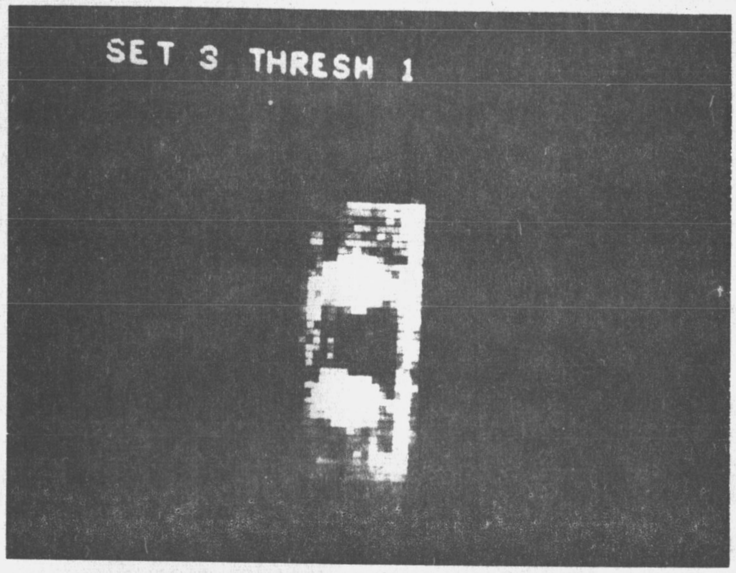
1. Baily, N.A., Three-Dimensional Image Reconstruction of Roentgenographic Tomograms by Holography, A.A.P.M. Summer School, Boulder, Colorado (1974).
2. Baily, N.A., Roentgenographic Storage Systems Through the Use of Holography, A.A.P.M. Summer School, Boulder, Colorado (1974).
3. Baily, N.A., Keller, R.A., Steigerwalt, J.E., and Crepeau, R.L., The Use of a Computerized Electronic Radiography System For Quantitative Measurements, 27th Annual Conf. on Eng. in Med. and Biol., Philadelphia, Pennsylvania (1974).

PAPERS PUBLISHED

1. Fluoroscopic Tomography, Baily, N.A., Crepeau, R.L., and Lasser, E.C., Invest. Radiol. 9, 94 (1974).
2. Performance of a Large Screen Fluoroscopic Imaging System, Baily, N.A. and Crepeau, R.L., Proc. of the S.P.I.E., 43, 135 (1974).

PAPER SUBMITTED FOR PUBLICATION

Baily, N.A. and Crepeau, R.L., Capabilities of Fluoroscopic Systems to Determine Differential Roentgen Ray Absorption, Radiology.



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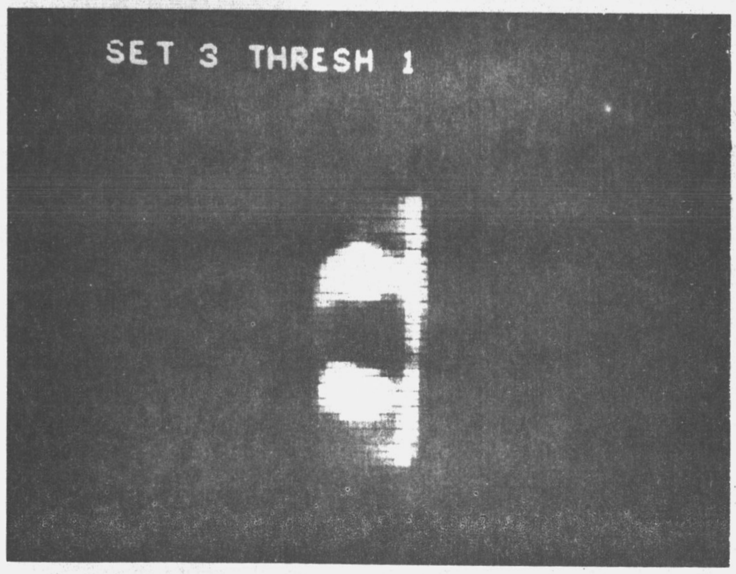

2

FIGURE 5

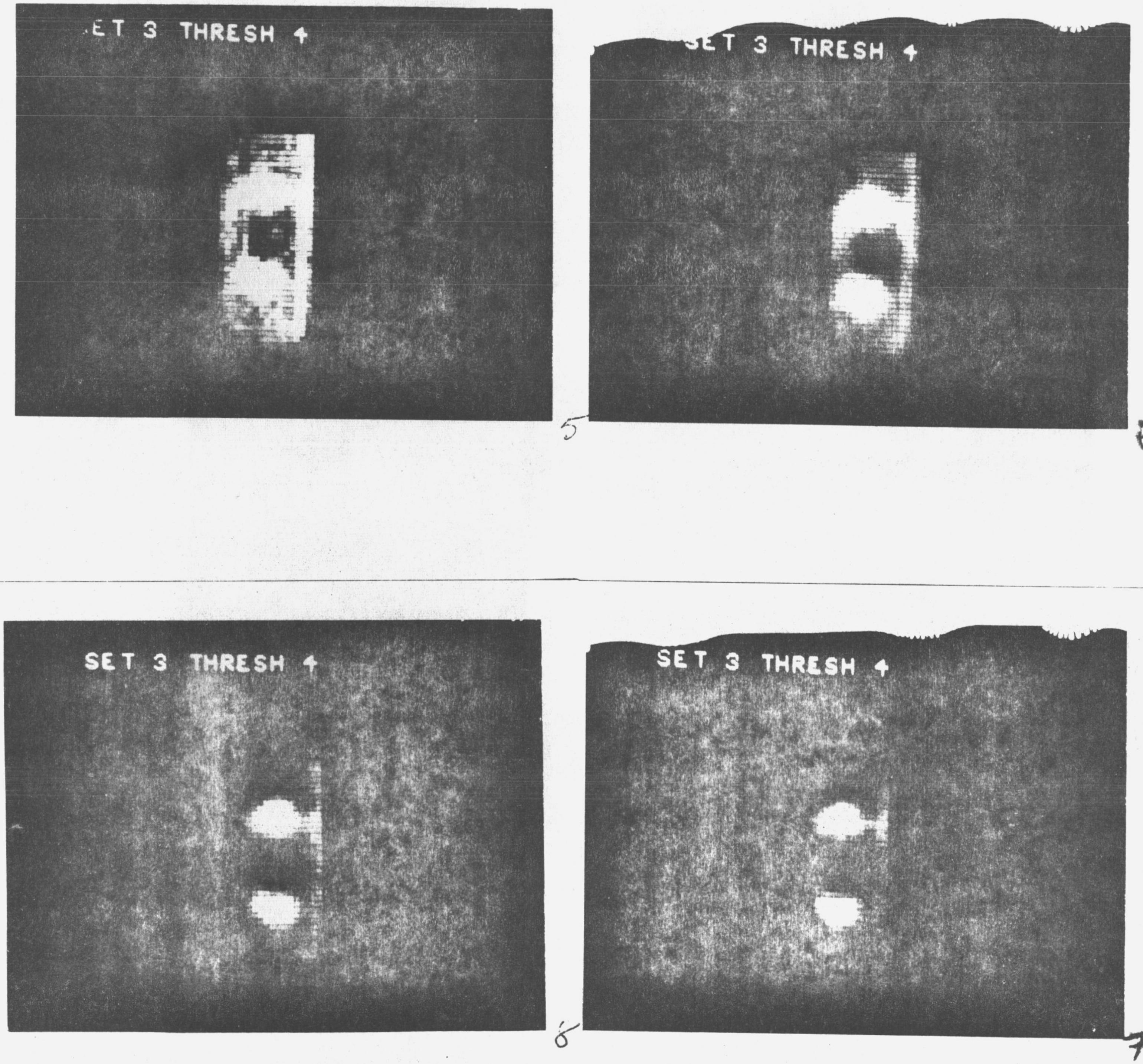

FIGURE 6

PERSONNEL PARTICIPATING IN PROGRAM

1. Norman A. Baily, Professor of Radiology
2. Elliott C. Lasser, Professor of Radiology
3. Avinash Kak, Associate Professor of Electrical Engineering (Purdue Univ.)
4. Robert A. Keller, Associate Specialist, Radiology
5. Earl M. Raeburn, Lab Technician
6. Richard J. Nachazel, Lab Technician